Abstract - Research suggests that fluctuations in diurnal IOP are a significant independent risk for glaucoma progression. An effective self-tonometer and a clinical method for its use by the patient at home would help the management of glaucoma. We describe a prototype of a new tonometer that may prove useful for non-invasive self-tonometry. Preliminary data suggests that the device is locating the area of IOP by calculating the force required to deform the cornea at a given time.

I. INTRODUCTION:
Elevated intraocular pressure (IOP) has been associated with visual field loss in glaucomatous patients [1]. Presently, to acquire this measurement, the clinician must anesthetize the eye and take the measurement directly from the cornea. Most patients with glaucoma have their IOP taken by the clinician about every two months. According to the patient’s IOP, the clinician prescribes the appropriate medication protocol that will reduce the IOP. However, it is also known that there are diurnal variations in IOP that have been found to be the highest in the early morning just after waking but resolve within an hour [2]. Currently, diurnal variation in IOP is not monitored daily which may cause the patient to have acceleration visual field loss. The diurnal variation results in a less reliable prognosis by the clinician and facilitates the necessity for a noninvasive self-operated tonometer that is affordable and easily operated by the patient at home.

II. METHODOLOGY:
A micro strain gauge, which accurately measures small forces from 0-50 grams, was integrated with an unguided linear displacement transducer (LVDT), which is used to measure small precise changes in distance within the range of 0-0.1 inches. The combination of these two transducers, as well as, a circuit for signal conditioning and a custom LabVIEW program make it possible for both signals to be time synchronized, saved and analyzed.

The incorporated design has a custom tip that is applied to a closed eyelid directly in front of the cornea (Fig 1). The force is gently increased by the clinician or patient until necessary corneal deformation takes place; meanwhile, the LabVIEW program is processing and compiling the data at a sampling rate of 1000 samples per second. The stream of data that is sampled and collected is sent from LabVIEW to an Excel file. The force measurement and distance measurements are synchronized. A conversion is done to change the force voltages to grams using the calibration data and then to mmHg by use of the following conversion equation.

\[
M(g) \times \left( \frac{0.001 (kg/g)}{\pi (0.003)^2 (m^2)} \right) \times \left( \frac{9.8 (m/s^2)}{} \right) \times \left( \frac{1 (mmHg)}{133.3 (kg*m^3*s^{-2})} \right) = M (g) \times (2.60) mmHg / g
\]

The schematic (Fig.2) shows the combination of the transducers with respective signal conditioning components and the data interface components. The 10 V dc power supply feeds the micro strain gauge (Fig.2-A) which, in turn, outputs a small dc signal of approximately 0-15 millivolts dc depending on the applied pressure. The signal is then amplified through the op-amp (Fig.2-B) to produce a dc signal from 0-10 V dc. The amplified signal is filtered by a passive low pass filter (Fig.2-C). The signal is fed into a breakout box, which delivers it to the DAQ card that interfaces the signal to the custom LabVIEW program (Fig.2-D). The Function generator feeds a 5 Vrms @ 5 kHz sinusoidal signal to the LVDT (Fig.2-E). The 0-
1 Volts rms signal, depending on position, is amplified by the op-amp in (Fig.2-F). The amplified, approximately 0-10 Volts rms, signal is rectified by the full wave bridge (Fig.2-G) and filtered to a DC signal by a passive low pass filter (Fig.2-H). This resulting signal is also introduced into the breakout box, DAQ card and custom LabVIEW program (Fig.2-D) as described above.

A calibration of both transducers was completed prior to taking the data. The LVDT was fixed and a spring was loaded to apply a constant outward force on the LVDT armature. A micrometer was fixed in a head on position to the armature so that signal output measurements could be recorded with precision down to 0.001 of an inch. A linear range was established and a non-loaded stop was designed and built to allow the LVDT to stay within this range. The calibration linearity for the LVDT can be seen in Figure 2. The micro strain gauge was fixed on a precision digital balance whose resolution was 0.01 grams. A micrometer was fixed above the micro strain gauge so that a constant force could be applied with a resulting DC signal. A mechanical stop was also designed and built to allow the micro strain gauge to only operate within its mechanical limits. The calibration linearity for the micro strain gauge is shown in Figure 3.

![Strain Gauge Linearity](image)

**Figure 3: LVDT Linearity**

\[ y = 0.0171x + 0.1563 \]

\[ R^2 = 1 \]

![LVDT-Linearity](image)

**Figure 4: Strain Gauge Linearity**

III. RESULTS:

Preliminary graphs show good linear characteristics; however, as the pressure at the tip is increased to at or above the IOP, the cornea is deformed. There is a reduction in force over a given distance which causes a trough like area on the graph. Since the trough represents the area where the tip moved without an increase in pressure, this trough denotes the area of IOP. The subject’s trough fell in between 18 and 20 mmHg (Fig.5). The subject had their IOP taken with the gold standard for clinical IOP determination, the Goldman tonometer, and the IOP measurement was 18 mmHg; however, some graphs showed lower troughs in Figure 5.

IV. DISCUSSION:

More subjects will need to be incorporated into the study to make a better representation of the data. These subjects will need a baseline Goldman reading for comparison. This information may be necessary to make a correlation with a specific area of the perceived trough. Also, the larger data set may facilitate the necessity of an empirical correction factor to the modified Imbert-Fick law which is based on the equation \( W = P \times X \); where \( W \) = external force, \( P \) = pressure, and \( A \) = area. Correction factors are implemented to the Imbert-Fick equation to apply the law to the eye since the eye does not fit all parameters of the law.[3].

Most likely, tip size will play a large roll in error reduction and increased repeatability. The large tip is more susceptible to misalignment and; therefore, distributing the force over a smaller and unknown area of the tip.[3,4]. Thus, making the conversion from grams to force inaccurate. A smaller tip will reduce the amount of error produced by misalignment and hopefully translate into greater reproducibility.

A prototype for a noninvasive self operated tonometer displayed the necessary characteristics theorized, such as the, force vs. distance relationship over time and a definite inflection point after an otherwise linear relationship to justify further investigation as a medical device.

ACKNOWLEDGEMENT: State Commission Grant

REFERENCES